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(54) Title: METHOD AND APPARATUS FOR DISPLAYING INFORMATION OBTAINED BY ELECTRICAL IMPEDANCE TOMOGRAPHY DATA
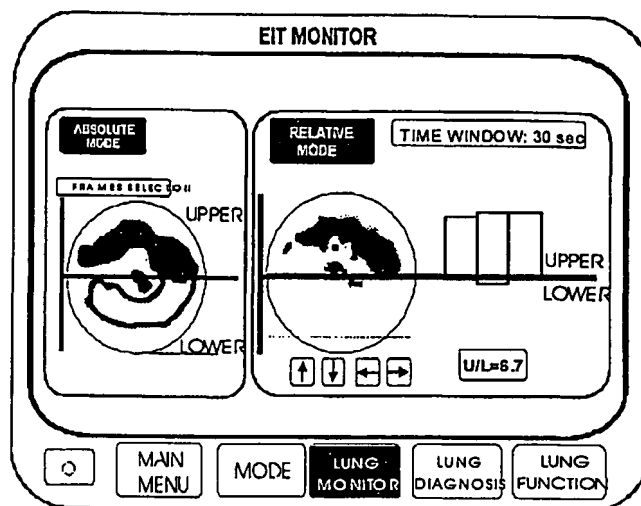
(57) Abstract: The invention refers to a method and an apparatus for displaying information obtained by electrical impedance tomography (EIT) data from a part of a patient's body. In order to provide a method for displaying information which can be im-mediately used in order to determine a specific pathological condition of said part, a method according to the invention performs the following steps: processing the EIT data with a plurality of predetermined processing modes; determining specific pathological con-ditions of said part in accordance with the results of the predetermined processing modes; selecting a screen mode from a plurality of predetermined screen modes in accordance with the specific pathological conditions; and displaying the EIT data in accordance with the selected screen modes.

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Method and Apparatus for Displaying Information
obtained by Electrical Impedance Tomography Data

The invention refers to a method and an apparatus for
displaying information obtained by electrical impedance
tomography (EIT) data from a part of a patient's body.

The electrical impedance tomography takes advantage of the
differing specific conductivity of human tissues, which
varies from 15.4 mS/cm for cerebrospinal fluid to 0.06 mS/cm
for bone. The difference in the value of conductivity is
large between different tissues. Cross sectional images of
the distribution of conductivity or alternatively specific
resistance therefore show a good contrast. The aim of
electrical impedance tomography is to produce images of those
contrasts.

An example of carrying out an EIT measurement is the analysis
of a patient's lung. A number of electrodes are placed around
the thorax, wherein an alternating current with e.g. 50 kHz
at 5 nA peak to peak amplitude is applied to respectively
adjacent electrodes. The other electrodes respectively are
used with the alternating current to carry out the
measurement of impedance against a defined reference
potential. As soon as all the electrodes, one after another,
have served as current conducting electrodes, a cycle for
data detection is concluded. In order to eliminate
statistical disturbances, as a rule a plurality of data

detection cycles is averaged, in order to obtain a
corresponding picture.

The maximum impedance changes in the zone of the thorax are
caused by breathing in and out of air. In this context it can
be observed that the impedance change which is measured by
electrodes is a measure of the change of volume in the lung.
Therefore, according to the process of EIT, measurements can
also be carried out with respect to the pressure-volume
relationship of the lung.

The complete reconstruction problem is non-linear and
requires iteration. However, each step in the iterative
process is linear. Images reconstructed using only the first
step of iteration effectively treat image formation as a
linear process, an assumption approximately justified for
small changes in conductivity from uniform. Most of the
clinical images produced today are using a single-step
reconstruction algorithm.

One aim of EIT is to reconstruct images of the absolute
distribution of conductivity. These images are known as
absolute images. However, this requires that the forward
problem can be solved to a high degree of accuracy, and this
can be difficult. The magnitude of the voltage signal
measured on an electrode or between electrodes will depend on
the body shape, the electrode shape and position, and the
internal conductivity distribution. The signal magnitude is
in fact dominated by the first two effects rather than by
conductivity. However, if a change in conductivity occurs
within the object, than it can often be assumed that the
change in surface voltage is dominated by this conductivity

change. In differential imaging, the aim is to image changes
in conductivity rather than absolute values.

Differential algorithms can only image changes in
conductivity. Absolute distributions of conductivity cannot
be produced using these methods. In addition, any cross
movement of the electrodes, either because they have to be
removed and replaced or even because of significant patient
movement, make the use of this technique difficult for long-
term measurement of changes. As an alternative to changes in
time, differential algorithms can images changes in
conductivity with frequency. Measurements can be made over a
range of frequencies and differential images can be produced
using data from the lowest frequency and the other
frequencies in term. A multi-frequency measurement thereby
makes use of the complex resistance of a tissue which depends
on the frequency.

As it becomes obvious, the analysis of a patient's lung by
electrical impedance tomography yields a vast amount of data.
An EIT image consists of a plurality of pixels, wherein each
pixel can be determined by different reconstruction
techniques as described above, i.e. by the determination of
the absolute distribution, the relative distribution or
distribution over a range of frequencies.

On the other hand, there are also a plurality of lung
conditions which have to be determined from the plurality of
EIT data. Basically, this derives from the fact that in the
lung there are theoretically four types of alveoli, which is
shown in Fig. 1. The normal alveolus (A) is both ventilated
and perfused with blood. There are alveoli that are
ventilated, but not perfused (B); such alveoli contribute

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significantly to the physiologic dead space. There are
alveoli that are not ventilated, but perfused (C); such
alveoli do not provide the exchange of respiratory gases.
Finally, there are alveoli that are both poorly ventilated
and poorly perfused (D); such alveoli contain high CO_2 and N_2
and O_2. These alveoli are the last to expel their CO_2 and N_2
in washout tests.

An experienced doctor is able to use the plurality of EIT
data in order to determine the plurality of different lung
conditions. This is conventionally done by analysing the
different types of reconstruction images according to the
absolute distribution, the relative distribution and the
distribution over a certain range of frequency. However, even
for an experienced doctor it is still very time-consuming
using a conventional EIT monitor to come to sufficient
results.

Therefore, it is an object of the invention to provide a
method for displaying information obtained by electrical
impedance tomography data from a part of a patient's body
which can be immediately used in order to determine the
respective pathological condition of said part.

This object is solved by a method according to claim 1 and an
apparatus according to claim 21. The inventive method
comprises the steps of:

-    processing the EIT data with a plurality of
     predetermined processing modes;

- determining specific pathological conditions in
  accordance with the results of the predetermined
  processing modes;

- selecting at least one screen mode from a plurality of
  predetermined screen modes in accordance with the
  specific lung conditions; and

- displaying the EIT data in accordance with the selected
  screen modes.

In the following, six predetermined processing modes
according to the invention are explained in the case of
analysing a patient's lung. These modes are

- the relative mode,
- the phase-lag mode,
- the perfusion mode,
- the absolute mode,
- the time constant mode and
- the regional spirometry mode.

Relative Mode

The relative mode calculates regional changes of a bi-
dimensional distribution of tidal ventilation during a time
period in the past. The regional changes in impedance are
obtained off-line by electric impedance tomography at
discrete time intervals utilizing a technique capable of
estimating relative variations in impedance (percent
variations, for instance). Relative variations in impedance
are calculated by using a simplified model of the region,
mathematically representing the cross sectional plane of the
thorax containing the electrodes. The simplified model is

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obtained through a variety of methods, like the finite
element method, or finite differences, or even Eigen-function
superposition. Computation of relative changes in impedance
implies the choice of a reference. This reference may be a
set of boundary voltages or a vector of impedance
distributions in the region of interest, obtained in a past
moment or during a time interval in the past. This particular
moment or interval of reference is chosen manually or
automatically. After the completion of a certain time period,
a sequence of relative impedance changes for each pixel is
available, and some measurement of its variance along the
time is calculated. The variance estimation is a statistical
measure of dispersion, like amplitude, standard deviation, or
others.

Phase-Lag Mode

The phase-lag mode processes the dynamics of tidal
ventilation and is an index of the homogeneity of the lung
dynamics. Values can be displayed for inspiration, expiration
or for the whole respiratory cycle. Narrow ranges represent
the typical homogeneous behaviour of lungs with healthy
airways and alveoli. Larger ranges are indicative of severe
airway and alveolar disease.

Fig. 2 shows the processing of EIT data according to the
phase-lag mode.

During monotonous tidal ventilation, the phase angle of each
pixel is calculated when compared to the sinusoidal or some
other approximating function of the overall impedance
oscillation during tidal breaths. By considering all pixels
representing a lung area, the overall impedance oscillation
represents the average impedance oscillation. The individual

pixel oscillations will be compared to this overall oscillation for phase lag detection. The overall impedance oscillation, as well as the individual pixel oscillation, is calculated by relative electric impedance tomography at discrete time intervals, as described above. A negative phase angle means that a certain pixel is being inflated/deflated earlier than the average lung oscillation, whereas a positive phase angle means that the pixel represents a lung zone with a delayed inflation/deflation pattern. Each pixel has a phase angle representing it.

Perfusion Mode

The perfusion mode processes the bi-dimensional distribution of lung perfusion during a time period in the past. The same principles used for the relative mode (described above) are applied here. The only particular aspects are as follows: The sampling rate has to be > 20 Hertz. Cardiac related changes in resistivity in the lungs are smaller (about 5 - 10% of those related to breathing movements) and occur at an oscillatory frequency ten times higher. The whole cycle of lung perfusion oscillation has precisely the length of the electrical cardiac cycle (usually in the range of 0.5 to 1 sec). After the completion of a certain window of time, a sequence of cardiac-related-relative-impedance-changes is available and some measurement of its variance along the time is calculated off-line. The variance estimation can be any statistical measure of dispersion like standard deviation, variance per se or others. The pixel values will be proportional to this estimate of dispersion (for each pixel) during the time period in question.

To improve the quality of the perfusion signals of EIT an apnea period is introduced. During apnea, we assume that the

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dispersion measure is proportional to the regional perfusion. Alternatively, the patient can be ventilated at a respiratory rate well below (less than 1/3) the cardiac rate, and a high-pass filter can be used to remove breath related changes in impedance.

Considering a fixed cardiac output, the amplitudes of impedance oscillations along the cardiac cycle are proportional to the relative blood perfusion into a certain lung region. Considering two equally sized lung regions or pixels, the one showing the highest oscillation related to cardiac cycles is the one receiving a greater proportion of the cardiac output.

In order to increase the amount of information obtained from a lung perfusion EIT image even further, it is advisable to subdivide the entire cardiac cycle into its components. As an example, consider a patient with a cardiac rate of 60 beats/min and a sampling rate of 50 Hertz. The hypothetical collection of subsequent sets of impedance distributions, at equally timed intervals, starting from the peak of QRS (the ventricular related electrical depolarisation detected on the ECG) and finishing 20 msec before the next QRS complex will reveal 3 waves of impedance changes:

First, a small fall in impedance is related to the contraction of the left atrium pushing blood back into the lungs. A second big fall in impedance is related to the stroke volume delivered to the pulmonary arteries by the right ventricle; and finally a third intermediate fall is due to the elastic recovery of the energy stored in pulmonary arteries, pushing blood through the pulmonary capillaries and veins. The whole cycle will be repeated, starting from the

next QRS complex. An analysis of each of these sub-phases of
the cardiac cycle can be performed separately using the same
display functions as described previously for the other
modes.

Absolute Mode

The absolute mode processes the estimated absolute impedance
of each pixel, after correction for noise and electrode/skin
impedance problems. Different from the algorithms for
calculating relative changes in impedance, iterative
mathematical algorithms are applied for stepwise convergence
to the right solution (defined, for instance, as the
impedance distribution in a ROI, generating the lowest
possible squared error between the predicted boundary
voltages and those actually measured). The convergence may be
guided by some noise analysis. The pixel values represent an
estimate of absolute impedance for each pixel – the average
value during the data collection time.

As described above, the absolute mode processes the estimated
absolute impedance of each pixel. Relative changes in
impedance are computed directly by using one step methods;
absolute impedance measurements, however, are obtained by
iterative mathematical algorithms. The best estimated
distribution is always an impedance distribution on the ROI
that minimizes a certain error index. The error index may be
defined in different ways, according to the observed
differences between the predicted boundary voltages and those
actually measured. The algorithm convergence may be guided by
some noise analysis, as in the Kalman filter algorithm.
Several procedures may be used to filter noise as singular
value truncation or coherent means of measured voltages.

To achieve the best quality of data, the patient has to be on
static conditions during the data acquisition time, or the
data acquisition interval shall be short relatively to the
respiratory cycle. Alternatively, a long time period of data
acquisition can be used and some low-pass filtering process
can be used.

In order to enhance the quality of the information obtained,
we suggest to use more than one algorithm for the estimation
of impedance distribution together, having two or more
calculating procedures working in parallel. A slower but more
precise algorithm can be used as a check procedure for the
faster one. Genetic algorithms, neural nets, Newton Raphson
methods or Kalman-Filter-based-algorithms can be used.
Ambiguities in the solution of a fast algorithm can be solved
by genetic algorithms, which progressively improve their
performance along the time. Neural nets, previously trained
with data collected from conventional CT or bedside clinical
data, can improve the definition of common structures inside
the thorax.

Once a sequence in time of absolute impedance distributions
is computed, the relative impedance distribution can be
obtained by algebraically subtracting two subsequent absolute
distributions. Relative impedance changes computed by the
non-linear models mentioned above are a better estimate of
the true changes than the ones calculated by traditional back
projection techniques.

In the absolute mode, aerated zones will be assigned with a
high pixel value, whereas liquid/solid zones will be assigned
with a low impedance value, independent of its variation
along tidal breathing. Bubbles of trapped air (pneumothorax

or bubbles of emphysema, for instance) will be represented by
high impedance values, despite the fact that they would be
assigned low values for variance/dispersion in the relative
mode.

Because absolute impedance can be defined as an absolute
number, with a dimension like Volts/Amperes/Pixel-Volume, the
colours, tonalities, or elevations can have an absolute
scale. Therefore, the overall colour of the lung inside the
region of interest will provide immediate information about
the degree of lung aeration/edema. Appropriate corrections
for the thoracic circumference and subcutaneous fat and
muscle can be provided with the help of an extensiometer or
plethysmograph.

Time Constant Mode

Fig. 3 shows the processing of EIT data according to the time
constant mode. During controlled mechanical ventilation or
during special manoeuvres in the awake and spontaneously
breathing patient (say a forced vital capacity manoeuvre) a
special exponential approximation function is used.

The exponential fit process will be applied for the overall
change of impedance (the whole region of interest), as well
as the regional (per pixel) impedance changes. By using an
exponential approximation during a relaxed expiration, the
overall time constant and the time constants for each pixel
can be estimated: i.e. the time period until a change
corresponding to 63.2% of the total impedance change during a
long exhalation (enough to achieve a stable plateau of
impedance). The start of the running window can be
synchronized with the beginning of exhalation by making use
of a manual or automatic trigger.

In this particular case, pixel values will represent time
constants displayed as colours or tonalities in a bi-
dimensional graphic display (or by elevation on the z-axis,
in a three-dimensional display). Because a time constant can
be defined as an absolute number - with its dimension of
seconds - the colours, tonalities, or elevations can have an
absolute scale. Therefore, the overall colour of the lung
inside the region of interest will provide immediate
information about the degree of lung obstruction. Zone
separations and ratio calculations can be performed.

Additionally, the dispersion of time constants can also be
used as an index of airway disease. Any of the above-
mentioned statistical measure of dispersion can be used for
this purpose.

Finally, the presence of units not achieving a stable plateau
of impedance, even after a long expiration, constitutes also
a signal of severe airway disease.

Regional Spirometry Mode
Spirometry is still one of the most important pulmonary
function tests. It is used to determine specifically defined
and standardized volumes - like the tidal volume - of inhaled
or exhaled breathing gases. By adding one or more of these
volumes lung capacities - like the vital capacity (VC) - are
determined. To date, a regional analysis of these volumes and
capacities has not been possible without invasive or
radioactive means.

An EIT monitor can enhance traditional spirometry by adding
information about the regional distribution of the gross

volumes within the lungs. In analogy with traditional spirometry, total but also regional volumes and capacities can be obtained by EIT. Time dependent volumes, like the FEV_1 (forced expiratory volume within the first second of exhalation) can be determined on a regional as well as on a global basis. When observing volume changes versus time, a flow can be calculated and analysed for the entire lung, but also for a smaller region of interest within the lung. This way, pathologies with respect to compliance and resistance are detected not only on a global, but also on a regional lung level and the regional predominance of such a pathology can be identified.

Each pathological condition or each lung condition respectively can be determined for the whole lung or for a region of interest of the lung. In the latter case lung conditions with local spreading can be determined more precisely. For example, the patients lung can be divided up into four regions of interest, wherein for each region of interest a full determination of each lung condition by running all respective processing modes is performed. As a result, for each of the region of interest the respective lung conditions are determined wherein the determined lung conditions can differ between the several regions of interest.

According to one aspect of the invention, the determination of specific lung conditions in accordance with the results of the predetermined processing modes can be based on a knowledge based data bank which contains certain decision rules for each specific lung condition. The decision rules can be based, e.g. on Fuzzy Logic.

As to the different types of screen modes, according to the
invention the following types of screen modes can be
distinguished in the case of analysing a patient's lung:

- functional images,
- graphs,
- numerical values and
- others.

Functional Images

A functional image is an online imaging of the patient's lung
on the basis of an actual processing mode. Thus, functional
images form a screen mode which are a graphic representation
of the underlying data for the automatic determination of the
specific lung conditions. Each pixel is displayed according
to its distribution and is preferably averaged over time by
applying a running window of e.g. 5 - 30 seconds. The
location of each pixel in the functional image presents the
estimate of its spatial location inside the thorax,
considering the plane embracing all electrodes.
Alternatively, a pixel value can be represented by its
evaluation on a third axis, using a three-dimensional
display.

Taking for example the relative or differential distribution
of conductivity, during the running window of time the
regional changes of electric impedance in each pixel within
the two-dimensional plane embracing all the electrodes are
allocated. After the completion of a certain window of time,
a sequence of relative impedance changes for each pixel is
available and some measurement of its variance along the time
can be calculated. Pixel values can be represented by colours
or tonalities in a two-dimensional graphic display.

All in all, the following distributions preferably can be
displayed:

-       absolute impedance,
-       relative impedance,
-       lung perfusion and
-       phase-lag.

Furthermore, besides these basic modes of functional images,
pixel values in the graphic display can also be determined by
any mathematical and/or logical operation between the
different basic modes. This leads e.g. to
ventilation/perfusion, dispersion over time constants and/or
image subtractions from the comparison of two modes.

As it can be noted the basic modes of functional images
correspond to the four processing modes as described above.
As a general rule, it can be assumed that the corresponding
functional image of a processing mode is always displayed, if
the corresponding processing mode leads to the determination
of a bad lung condition. On the other hand, if a healthy lung
condition is determined, it is advisable to further reduce
the amount of data so that not the basic functional images,
but appropriate graphs or numerical values are displayed.

Graphs
A graph is in principle a further data reduction with regard
to a functional image and is preferably selected, if a
healthy lung condition is determined. In principle, a graph
is an online display of measured or calculated data
representing selected regions of interest. Any form of graph
can be chosen, like a line graph or a graphic bar.

Preferably, there are the following representations of
graphs:

-    tidal ventilation of total lung,
-    tidal ventilation within a region of interest,
-    tidal ventilation within a plurality of lung regions,
-    ratio of ventilation within two regions of interest,
-    mean phase-lag of total lung,
-    mean phase-lag within a region of interest,
-    mean phase-lag within a plurality of lung regions,
-    ratio of ventilation within two regions of interest,
-    exponential curve for forced vital capacity manoeuvres
     as during lung function tests for the total lung and for
     any region of interest.

Numerical Values

Numerical values are a further data reduction with regard to
graphs or functional images. A numerical value is an online
display of measured or calculated values representing
condensed information on the entire lung or selected regions
of interest. As it is applied for graphs, numerical values
will be chosen as a selected screen mode, if a healthy lung
condition is determined in accordance with the results of the
predetermined processing modes.

Preferably, the following numerical values are assumed:

-    homogeneity index for ventilation within the total lung,
-    homogeneity index for ventilation within a region of
     interest,
-    ratio of ventilation in used impedance changes in
     selected parts of the lung, e.g. ventilation in used

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impedance changes of the upper lung divided by the lower
lung,

- homogeneity index for perfusion within the total lung,

- homogeneity index for perfusion within a region of
interest,

- ratios of perfusion in selected parts of the lung like
the upper/lower ratio,

- homogeneity index for phase-lag within the total lung,

- homogeneity index for phase-lag within a region of
interest,

- ratios of perfusion in selected parts of the lung like
the upper/lower ratio,

- numerical value or a time constant for characterizing a
forced vital capacity manoeuvre as during lung function
tests for the total lung and/or for any region of
interest.

Others

Of course, any other suitable screen mode is possible for
displaying the EIT data appropriately. One possibility is an
adapted decision support display which can be selected and/or
adapted in advance. One form of an adapted decision support
display can be a display for making clinical titration of the
positive end-expiratory pressure (PEEP) easy. The display can
consist of a coloured vertical or horizontal bar. The colours
range symmetrically from green in the very centre via yellow
and orange to dark red at both ends of the bar. Red indicates
too high or too low levels of PEEP, whereas green indicates
the optimal PEEP range. Any other combination of colours or
shading of one single colour may be used, too. A marker
within the bar indicates the current level of PEEP. Too high
levels of PEEP will place the marker into the upper or the
right red field, wherein too low levels of PEEP will move the

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marker downward or to the left. When operating the mechanical
ventilator, the user will just watch the marker move within
the bar. Once the marker has reached the middle of the green
zone, PEEP should have reached its optimal level.

In accordance with one aspect of the invention, the screen
mode is updated on demand of a user. This means that the
display of the selected is frozen until the user requests
another update of the underlying data of the display.

In accordance with another aspect of the invention, the
displayed EIT data are updated in real time. This means that
each display uses the data which are actually acquired from
the patient's body. If necessary, a running time window can
be used in order to achieve a further noise reduction.

In accordance with another aspect of the present invention,
an estimate of a pathological condition is input by a user,
wherein processing of the EIT data is carried out with a
subcombination of the predetermined processing modes in
accordance with the estimated pathological condition. In case
when the doctor has already an idea of the pathological
condition of the examined part of the patient's body, it can
be taken advantage of this knowledge by providing a
corresponding input format.

In accordance with another aspect of the present invention,
said part of the patient's body to be examined is the
patient's lung. Using the method of electrical impedance
tomography, it has shown very advantageous to examine a
patient's lung because very high impedance changes in the
zone of the thorax are caused by breathing in and out of air
so that the zone of the thorax shows good contrasts. Using as

processing modes the above-described relative mode, phase-lag
mode, perfusion mode and absolute mode, the following logic
can be applied for determining specific lung conditions:

The lung condition of a pulmonary infarct is determined, if
the relative mode yields a low result, the perfusion mode
yields a low result and the absolute mode yields a low
result.

The lung condition of a pulmonary embolism is determined, if
the relative mode yields a normal result, the perfusion mode
yields a low result and the absolute mode yields a normal
result.

The lung condition of a localized pneumonia is determined, if
the relative mode yields a low result, the perfusion yields a
high result and the absolute mode yields a low result.

The lung condition of an emphysema-like area is determined,
if the relative mode yields a low result, the perfusion
yields a low result and the absolute mode yields a high
result.

The lung condition of a cyclic airway closure is determined,
if the relative mode yields a high result, the phase-lag mode
yields a very high result and the absolute mode yields an
unsteady result.

The lung condition of persistent airway closure with air-
trapping is determined, if the relative mode yields a very
low result, the phase-lag mode yields a very high result and
the absolute mode yields an unsteady result.

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The lung condition of a persistent airway closure with re-
absorption atelaktasis is determined, if the relative mode
yields a very low result, the phase-lag mode yields an
unsteady result and the absolute mode yields a very low
result.

The above-described logic to determine specific lung
conditions can be based on the principles of Fuzzy Logic.
Each term of "low" and "high" has to be described in certain
numerical ranges with regard to the respective processing
mode. This can be done, e.g., by examining a patient with a
healthy lung and determining the numerical ranges for each
processing mode which can be applied for healthy conditions.
"Low" and "high" deviations from these reference values can
also be expressed by certain ranges of values.

According to a further aspect of the present invention, a
screen mode consists of graphic elements which are pre-stored
in a data store as graphic patterns and which are fetched
from the data store as soon as the corresponding screen mode
is selected. Thereby a faster graphic display can be achieved
because the graphic elements which involve time-consuming
build-up of the display are pre-stored already, wherein only
changes with regard to the actual EIT data have to be updated
in the pre-stored display.

The invention will now be described by way of an example and
with reference to the accompanying drawings in which:

Fig. 1   shows the four principle conditions of alveoli,

Fig. 2   shows the processing of EIT data according to the
           phase-lag mode,

Fig. 3   shows the processing of EIT data according to the
              time constant mode,

Fig. 4a   shows the determination of specific lung conditions
              in accordance with different processing modes,

Fig. 4b   shows the selection of screen modes in accordance
              with the specific lung conditions according to Fig.
              4a,

Fig. 5a   shows the determination of specific lung conditions
              in accordance with different processing modes,

Fig. 5b   shows the selection of screen modes in accordance
              with the specific lung conditions according to Fig.
              5a,

Fig. 6a   shows the determination of specific lung conditions
              in accordance with different processing modes,

Fig. 6b   shows the selection of screen modes in accordance
              with the specific lung conditions according to Fig.
              6a,

Fig. 7a   shows the determination of specific lung conditions
              in accordance with different processing modes,

Fig. 7b   shows the selection of screen modes in accordance
              with the specific lung conditions according to Fig.
              7a,

Fig. 8a   shows the determination of specific lung conditions
           in accordance with different processing modes,

Fig. 8b   shows the selection of screen modes in accordance
           with the specific lung conditions according to Fig.
           8a,

Fig. 9   shows the combined display of different screen
           modes,

Fig. 10   shows the combined display of another combination
           of screen modes

Fig. 11   shows a flow chart for performing the method
           according to the invention and

Fig. 12   shows the integration of an apparatus according to
           the invention in a closed-loop ventilation system.

As to the figures Fig. 1 to Fig. 3 it is referred to the
description above.

Figures 4 to 8 show the determination of specific lung
conditions in accordance with the results of predetermined
processing modes and consequently the selection of screen
modes in accordance with the determined specific lung
conditions. For the determination of the specific lung
conditions and the selection of the screen modes, a tabular
representation is used. Within each table field an indicator
shows the deviation from a normal result of a processing mode
or a normal range of a screen mode. An up-turned error means
a deviation to a value which is higher than the normal value,
whereas a down-turned error means the deviation from the

normal value by a lower value. Two or more errors indicate a strong deviation. Opposite errors indicate strong deviations around the normal value.

The lung conditions according to figures 4 to 7 refer to pathological conditions which usually affect only specific areas of the patient's lung. Therefore, as a overall principle according to the figures 4 to 7, the selected screen modes correspond to the processing modes, if the determined specific lung conditions are bad, i.e. not healthy. The reason for this is that in the case of a bad lung condition the doctor should have the same information available which led to the automatic determination of said lung condition in order to examine and approve the determined lung condition. However, if no bad lung condition is determined according to the figures 4 to 7, it is generally suggested to display overall graphs or overall numerical values which lead to a further data reduction. Examples of suitable graphs and numerical values are already listed above.

Another case is the determination of lung conditions which affect usually not a specific area of the lung, but the lung in total. In this case, the lung conditions can be characterized more specifically by adding the following values to the display:

- numerical value for the overall ventilation homogeneity,
- numerical value for the overall phase-lag homogeneity,
- numerical value for ratio of upper/lower lung ventilation – and used impedance variation and
- numerical value for ratio of upper/lower lung phase-lag.

Fig. 4 shows the determination of specific lung conditions on
the basis of three processing modes, namely the relative
mode, the perfusion and the absolute mode. This enables the
diagnosis of pulmonary infarct, pulmonary embolism, localized
pneumonia and/or emphysema-like area. As soon as a specific
lung condition is determined according to Fig. 4a, the
functional images according to Fig. 4b will be provided in
accordance with the determined specific lung conditions and
in accordance with the specified ranges according to Fig. 4b.

Further optimisation of the determination of those specific
lung conditions can be achieved by additionally using a
relative/perfusion mode (V/Q) which is derived from the
relative mode and the perfusion mode by calculating the
pixel-by-pixel ratio of both modes for each pixel. In this
way, one can achieve a better discrimination of the
conditions according to Fig. 4a. The corresponding
determination of the specific lung conditions is shown in
Fig. 5a. Again, the selection of the screen modes according
to Fig. 5b results in functional images which relate to the
corresponding processing modes.

Figures 6 and 7 show another example of the selection of
screen modes on the basis of another set of specific lung
conditions. More particular, the specific lung conditions are
pneumothorax, pleural effusion, atelektasis and over-
distension.

The combination of three modes (relative, absolute and
perfusion mode) enables the determination of pneumothorax or
pleural effusion; both pathologies are represented by zones
of low variance - so-called silent zones in the relative
mode. While a pneumothorax is characterized by a

circumscribed zone of high impedance values in the absolute
mode, a pleural effusion is associated with low values. The
image obtained in the absolute mode can be superimposed or
compared to the images obtained in the other modes. An
aerated area (high absolute values) extending outside the
zones of high relative dispersions (great tidal oscillation
of relative impedance) represents a collection of air
possibly due to barotraumas (destruction of lung tissue by
mechanical ventilation or the disease process). A condensed
area, with liquid-like absolute impedance extending outside
the zones of high tidal oscillation (in relative impedance)
represents a collection of fluid, most likely a pleural
effusion or a totally collapsed part of the lung.

Corresponding to the representation according to Fig. 5, Fig.
7 shows again the improvement of the processing with regard
to the specific lung conditions by adding a
relative/perfusion mode (V/Q), in which each pixel represents
the pixel-by-pixel ratio between the pixel value in the
relative mode and the corresponding pixel value in the
perfusion mode. However, for the above lung conditions, the
addition of the relative/perfusion mode does not add any
display information since an area of pneumothorax and
effusion shows neither ventilation nor perfusion.
Furthermore, the relative/perfusion does not help in
differentiating between atelaktasis and over-distension.
Therefore, in Fig. 7b only the functional images according to
the relative mode, the perfusion mode and the absolute mode
are suggested.

Finally, Fig. 8a shows the combination of the relative mode,
the phase-lag mode and the absolute mode for the
determination of patent airway, cyclic airway closure,

persistent airway closure with air trapping and persistent
airway closure with re-absorption atelaktasis.

Even if the relative mode shows a homogeneous distribution of
ventilation induced variance across the sectional plane of
the thorax, a significant phase-lag means that some problems
in the small airways may be present (small airway closure,
for instance, due to emphysema, COPD, surfactant deficiency,
or ARDS). The presence of patent airways all across the lung
parenchyma can only be guaranteed when both modes (the
relative and the phase-lag) present a homogenous lung
inflation/deflation. Therefore, the coincidence of a phase-
lag-delay within an area presenting a reasonable relative
variance of impedance will be used as a sign of cyclic airway
closure.

As to the selection of screen modes, it is suggested
according to Fig. 8b to display only the functional image
according to the relative mode and the phase-lag mode, but
not the functional image according to the absolute mode.

Fig. 9 shows an example of the selection of screen modes by
displaying the combination of the absolute mode and the
relative mode. Additionally, the same display may show
different information on different parts of the screen at the
same time like the following example:

-       a functional image of dispersion (e.g. the standard
        deviation of ventilation and used impedance changes in
        each pixel),

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-    a numerical value representing a global functionality
     (e.g. overall homogeneity index of the total lung's
     ventilation),

-    a graph representing calculated global or regional
     functionality (say ventilation and used impedance
     changes of the total lung),

-    a numerical value representing calculated regional
     functionality (e.g. ventilation- and used impedance
     changes of the upper lung divided by the lower lung).

Fig. 10 shows the combination of the functional image of the
absolute mode and a decision support display. In this case,
the decision support display is a graph of impedance changes
over time of the upper lung and the lower lung when
performing a PEEP-reduction (PEEP = Positive End-Expiratory
Pressure).

Fig. 11 shows a flow chart for performing the method
according to the invention. In step 1, the EIT monitor is
started, either by pressing a separate start button or by
simply switching on the EIT monitor. In step 2, predetermined
processing modes are run for determining specific lung
conditions. As described above, suitable processing modes are
the relative mode, the perfusion mode, the absolute mode and
the phase-lag mode. The determination of specific lung
conditions can be carried out as described according to the
figures 4a, 5a, 6a, 7a and 8a.

The result of step 2 is an evaluation of different lung
conditions for different regions of interest of the lung.
Hence, a plurality of determined lung conditions are

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available. It has to be further noted that the determination
of the lung conditions usually leads to both lung conditions
with severe problems and other lung conditions with a healthy
condition. Therefore, in step 3, a sorting of all determined
lung conditions with respect to their severity takes place,
in order to display the corresponding screen modes in
accordance with their importance. In step 4, a decision is
taken whether at least one bad lung condition was determined.
If yes, further processing takes place with step 5, otherwise
processing is continued with step 7. The reason for the
decision in step 4 is the fact that for a complete healthy
lung it is advisable to perform a further data reduction when
selecting screen modes, whereas in case of at least one bad
lung condition a selection of screen modes should take place
in accordance with the processing modes which led to the bad
result.

Consequently, in step 5, a selection of screen modes takes
place of functional images corresponding to the respective
processing modes, if necessary in certain regions of
interest. If necessary, then the user has got the possibility
to scroll through various screen modes in accordance with the
sorted sequence according to step 4.

On the other hand, if there are only healthy lung conditions
determined, a selection of screen modes for further data
reduction takes place in step 7. Suitable screen modes for
further data reduction are e.g. graphs, numerical values or
other adapted decision support displays. As already described
according to step 6, there is again a possibility in step 8
to scroll on demand through the display to the next screen in
accordance with the sorted sequence as sorted in step 4.

Fig. 12 shows the integration of an apparatus according to
the invention in a closed-loop ventilation system. It has to
be observed that the EIT monitor according to the invention
yields already all necessary data which can be used by a
ventilator for a closed-loop control. It might be also
possible to provide an input format for the doctor for
inputting limiting values with regard to the respiration
pressure for closing and opening the lung.

30
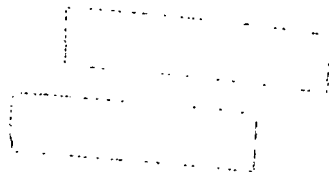
Claims

1. Method for displaying information obtained by electrical
    impedance tomography (EIT) data from a part of a
    patient's body, comprising:

    processing the EIT data with a plurality of predetermined
    processing modes;

    determining specific pathological conditions of said part
    in accordance with the results of the predetermined
    processing modes;

    selecting at least one screen mode from a plurality of
    predetermined screen modes in accordance with the
    specific pathological conditions; and

    displaying the EIT data in accordance with the selected
    screen modes.

2. Method according to claim 1, wherein the screen mode is
    updated on demand of a user.

3. Method according to claim 1, wherein the displayed EIT
    data are updated in real time.

4. Method according to one of the claims 1 - 3, wherein an
    estimate of a pathological condition is input by a user
    and wherein processing of the EIT data is carried out
    with a subcombination of the predetermined processing
    modes in accordance with the estimated lung condition.

5. Method according to one of the claims 1 - 4, wherein said part to be examined is a patient's lung.

6. Method according to claim 5, wherein a relative mode is implemented as a predetermined processing mode which yields as a result the mean variation of impedance changes due to ventilation.

7. Method according to one of the claims 5 - 6, wherein a phase-lag mode is implemented as a predetermined processing mode which yields as a result the mean phase lag of impedance changes with regard to ventilation changes.

8. Method according to one of the claims 5 - 7, wherein a perfusion mode is implemented as a predetermined processing mode which yields as a result the mean variation of impedance changes due to lung perfusion.

9. Method according to one of the claims 5 - 8, wherein an absolute mode is implemented as a predetermined processing mode which yields as a result the mean absolute impedance.

10. Method according to claims 6 and 8, wherein a perfusion ratio mode is implemented which calculates the ratio of the results of the relative mode and the perfusion mode and which yields as a result the ventilation/perfusion ratio.

11. Method according to one of the claims 5 - 10, wherein the EIT data are calculated on the basis of a running data window.

12. Method according to one of the claims 5 - 11, wherein the EIT data are calculated for a certain region of interest of the lung.

13. Method according to claim 12, wherein the lung condition of a pulmonary infarct is determined, if the relative mode yields a low result, the perfusion mode yields a low result and the absolute mode yields a low result.

14. Method according to claims 12 or 13, wherein the lung condition of a pulmonary embolism is determined, if the relative mode yields a normal result, the perfusion yields a low result and the absolute mode yields a normal result.

15. Method according to one of the claims 12 - 14, wherein the lung condition of a localized pneumonia is determined, if the relative mode yields a low result, the perfusion mode yields a high result and the absolute mode yields a low result.

16. Method according to one of the claims 12 - 15, wherein the lung condition of an emphysema-like area is determined, if the relative mode yields a low result, the perfusion mode yields a low result and the absolute mode yields a high result.

17. Method according to one of the claims 12 - 16, wherein the lung condition of a cyclic airway closure is determined, if the relative mode yields a high result, the phase-lag yields a very high result and the absolute mode yields an unsteady result.

18. Method according to one of the claims 12 - 17, wherein
    the lung condition of a persistent airway closure with
    air-trapping is determined, if the relative mode yields a
    very low result, the phase-lag mode yields a very high
    result and the absolute mode yields an unsteady result.

19. Method according to one of the claims 12 - 18, wherein
    the lung condition of a persistent airway closure with
    re-absorption atelectasis is determined, if the relative
    mode yields a very low result, the phase-lag mode yields
    an unsteady result and the absolute mode a very low
    result.

20. Method according to one of the claims 12 - 19, wherein a
    screen mode consists of graphic elements which are pre-
    stored in a data store as graphic patterns and which
    fetched from the data store as soon as the corresponding
    screen mode is selected.

21. Apparatus for displaying information obtained by
    electrical impedance tomography (EIT) data from a part of
    a patient's body, comprising:

    processing means for processing the EIT data with a
    plurality of predetermined processing modes and for
    determining specific pathological conditions of said part
    in accordance with the results of the predetermined
    processing modes;

    selecting means for selecting at least one screen mode
    from a plurality of predetermined screen modes in
    accordance with the specific pathological conditions; and

displaying means for displaying the EIT data in
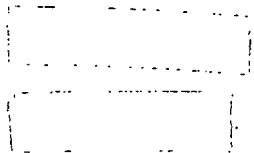accordance with the selected screen modes.

22. Apparatus according to claim 21, wherein the selecting
means is triggered on demand by a user for updating the
screen mode.

23. Apparatus according to claim 21, wherein the displayed
EIT data are updated in real time.

24. Apparatus according to one of the claims 21 - 23, wherein
an estimate of a pathological condition is input by a
user and wherein processing of the EIT data is carried
out with a subcombination of the predetermined processing
modes in accordance with the estimated lung condition.

25. Apparatus according to one of the claims 21 - 24, wherein
said part to be examined is a patient's lung.

26. Apparatus according to claim 25, wherein a relative mode
is implemented as a predetermined processing mode which
yields as a result the mean variation of impedance
changes due to ventilation.

27. Apparatus according to one of the claims 25 - 26, wherein
a phase-lag mode is implemented as a predetermined
processing mode which yields as a result the mean phase
lag of impedance changes with regard to ventilation
changes.

28. Apparatus according to one of the claims 25 - 27, wherein
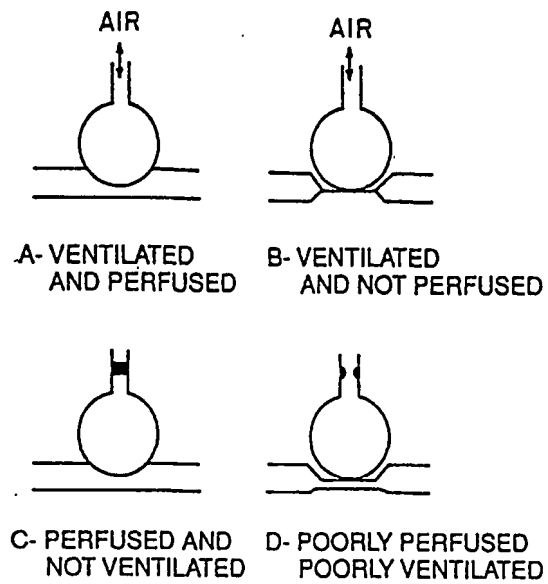a perfusion mode is implemented as a predetermined

processing mode which yields as a result the mean
variation of impedance changes due to lung perfusion.

29. Apparatus according to one of the claims 25 - 28, wherein
an absolute mode is implemented as a predetermined
processing mode which yields as a result the mean
absolute impedance.

30. Apparatus according to claims 26 and 28, wherein a
perfusion ratio mode is implemented which calculates the
ratio of the results of the relative mode and the
perfusion mode and which yields as a result the
ventilation/perfusion ratio.

31. Apparatus according to one of the claims 25 - 30, wherein
the EIT data are calculated on the basis of a running
data window.

32. Apparatus according to one of the claims 25 - 31, wherein
the EIT data are calculated for a certain region of
interest of the lung.

33. Apparatus according to claim 32, wherein the lung
condition of a pulmonary infarct is determined, if the
relative mode yields a low result, the perfusion mode
yields a low result and the absolute mode yields a low
result.

34. Apparatus according to claims 32 or 33, wherein the lung
condition of a pulmonary embolism is determined, if the
relative mode yields a normal result, the perfusion
yields a low result and the absolute mode yields a normal
result.

35. Apparatus according to one of the claims 32 - 34, wherein
    the lung condition of a localized pneumonia is
    determined, if the relative mode yields a low result, the
    perfusion mode yields a high result and the absolute mode
    yields a low result.

36. Apparatus according to one of the claims 32 - 35, wherein
    the lung condition of an emphysema-like area is
    determined, if the relative mode yields a low result, the
    perfusion mode yields a low result and the absolute mode
    yields a high result.

37. Apparatus according to one of the claims 32 - 36, wherein
    the lung condition of a cyclic airway closure is
    determined, if the relative mode yields a high result,
    the phase-lag yields a very high result and the absolute
    mode yields an unsteady result.

38. Apparatus according to one of the claims 32 - 37, wherein
    the lung condition of a persistent airway closure with
    air-trapping is determined, if the relative mode yields a
    very low result, the phase-lag mode yields a very high
    result and the absolute mode yields an unsteady result.

39. Apparatus according to one of the claims 32 - 38, wherein
    the lung condition of a persistent airway closure with
    re-absorption atelectasis is determined, if the relative
    mode yields a very low result, the phase-lag mode yields
    an unsteady result and the absolute mode a very low
    result.

40. Apparatus according to one of the claims 32 - 39, wherein a screen mode consists of graphic elements which are pre-stored in a data store as graphic patterns and which fetched from the data store as soon as the corresponding screen mode is selected.

Fig. 1
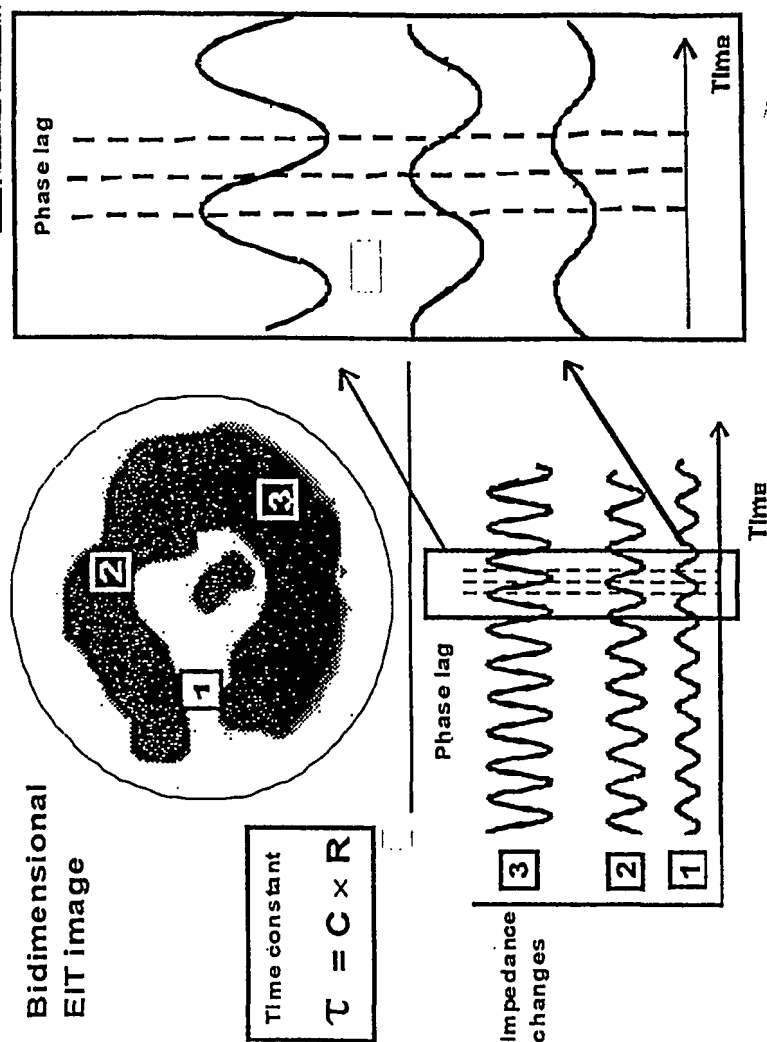
Fig. 2
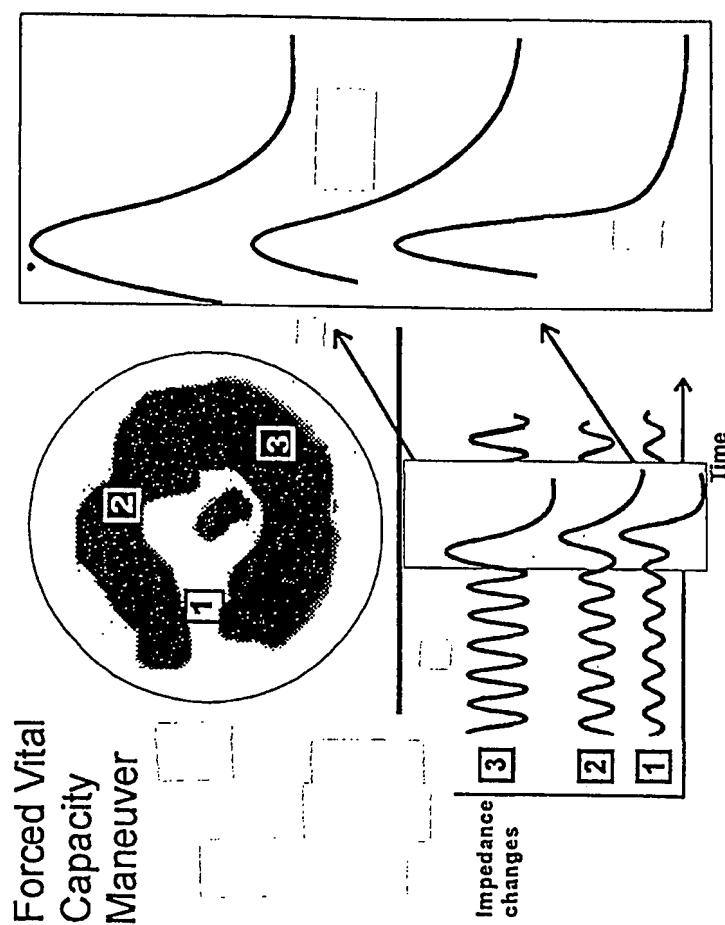
Forced Vital Capacity Maneuver

Impedance changes

Time

Fig. 3

processing mode / *Specific lung condition*	*relative mode*	*perfusion mode*	*absolute mode*
Pulmonary infarct	↓	↓	↓
Pulmonary embolism	Normal	↓	Normal
Localized pneumonia	↓	↑	↓
Emphysema-like area	↓	↓	↑

Fig. 4a

Screen mode / *Specific lung condition*	*functional image acc. relative mode in specific area*	*functional image acc. perfusion mode in specific area*	*functional image acc. absolute mode in specific area*
Pulmonary infarct	↓	↓	↓
Pulmonary embolism	Normal	↓	Normal
Localized pneumonia	↓	↑	↓
Emphysema-like area	↓	↓	↑

Fig. 4b

Specific lung condition \ processing mode	relative mode	perfusion mode	absolute mode	relative / perfusion mode (V/Q)
Pulmonary infarct	↓	↓	↓	↑
Pulmonary embolism	Normal	↓	Normal	↑↑↑
Localized pneumonia	↓	↑	↓	↓↓↓
Emphysema-like area	↓	↓	↑	↓↑

Fig. 5a

Specific lung condition \ Screen mode	functional image acc. relative mode in specific area	functional image acc. perfusion mode in specific area	functional image acc. abs. mode in specific area	functional image acc. V/Q in specific area
Pulmonary infarct	↓	↓	↓	↑
Pulmonary embolism	Normal	↓	Normal	↑↑↑
Localized pneumonia	↓	↑	↓	↓↓↓
Emphysema-like area	↓	↓	↑	↓↑

Fig. 5b

Specific lung condition \ processing mode	relative mode	perfusion mode	absolute mode
Pneumothorax	0	0	↑
Pleural effusion	0	0	↓
Atelectasis	↓	↓	↓
Overdistension	↓	↓	↑

Fig. 6a

Specific lung condition \ screen mode	functional image acc. relative mode in specific area	functional image acc. perfusion mode in specific area	functional image acc. absolute mode in specific area
Pneumothorax	0	0	↑
Pleural effusion	0	0	↓
Atelectasis	↓	↓	↓
Overdistension	↓	↓	↑

Fig. 6b

Specific lung condition \ _processing mode_	_relative mode_	_perfusion mode_	_absolute mode_	_relative/perfusion mode (V/Q)_
Pneumothorax	0	0	↑	Not available
Pleural effusion	0	0	↓	Not available
Atelectasis	↓	↓	↓	↓↓↓
Overdistension	↓	↓	↑	↓↓↓

Fig. 7a

Specific lung condition \ _Screen mode_	_functional image acc. relative mode in specific area_	_functional image acc. perfusion mode in specific area_	_functional sig. acc. absolute mode in specific area_
Pneumothorax	0	0	↑
Pleural effusion	0	0	↓
Atelectasis	↓	↓	↓
Overdistension	↓	↓	↑

Fig. 7b

processing *mode* / *Specific lung condition*	*relative mode*	*phase-lag mode*	*absolute mode*
Patent airway	Normal	Normal	Normal
Cyclic airway closure	↑	↑↑	↑↓
Persistent airway closure with air-trapping	↓↓	↑↑ (180°)	↑↓
Persistent airway closure with re-absorption atelectasis	↓↓	↑↓	↓↓

Fig. 8a

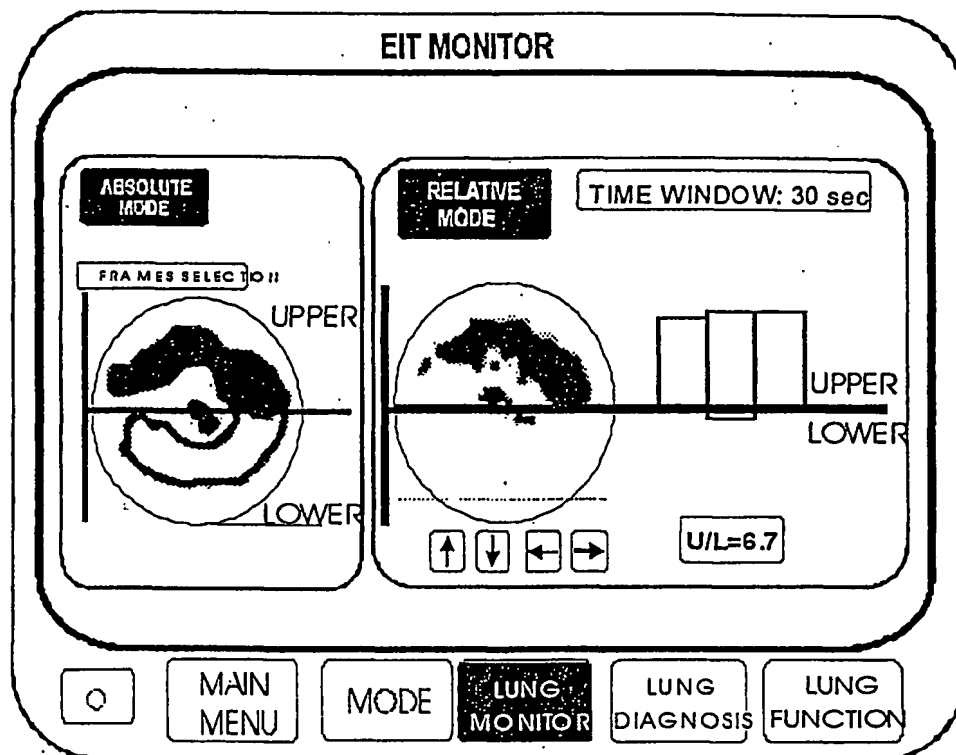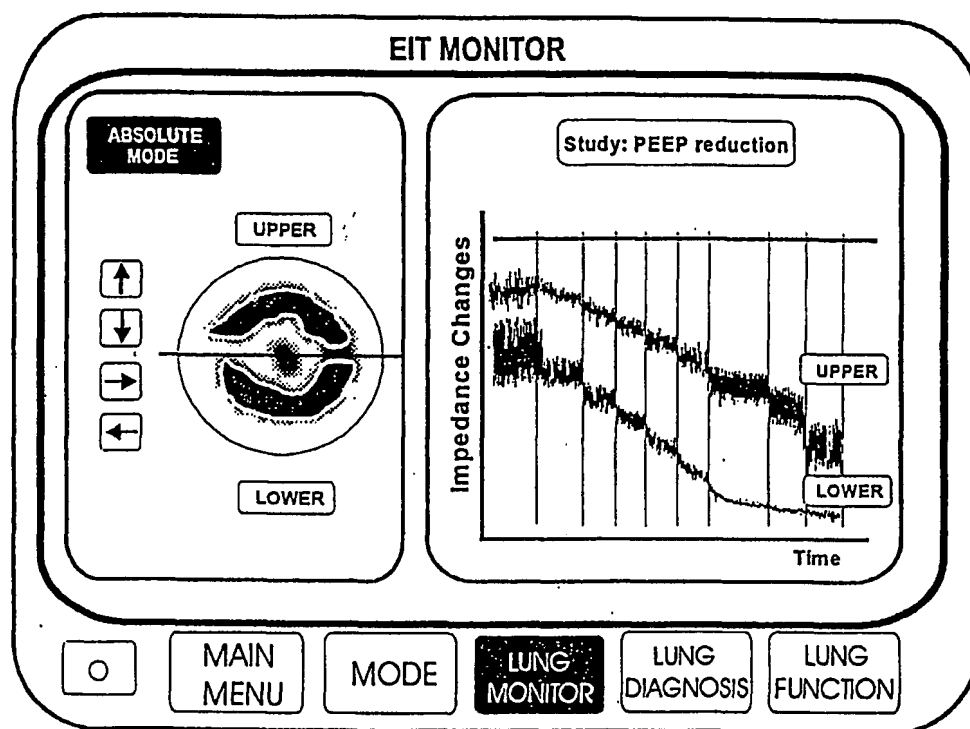| *screen* *mode* / *Specific lung condition*	*functional image acc. relative mode*	*functional image acc. phase-lag mode*
Patent airway	Normal	Normal
Cyclic airway closure	↑	↑↑
Persistent airway closure with air-trapping	↓↓	↑↑ (180°)
Persistent airway closure with re-absorption atelectasis	↓↓	↑↓

Fig. 8b

Fig. 9

Fig. 10

Start of the EiT-Monitor)~ 1

Running of processing modes and
determination of lung conditions ~ 2

Sorting the determined lung conditions
with respect to severity ~ 3

YES   At least
one bad lung condition   NO
was determined ~ 4

Selection of functional   Selection of Screen modes
images corresponding to   for further data reduction
5 ~ the respective
processing modes,   - graphs
if necessary in certain   - numerical values   ~ 7
regions of interest   - adopted decision support
                                   displays

←——— Scrolling on demand ———→

6 ~ Display of the next screen   Display of the next screen
in accordance with the   in accordance with the   ~ 8
sorted sequence   sorted sequence

Fig. 11

VENTILATOR

PATIENT

EIT MONITOR

COMPUTER SYSTEM

Fig. 12

INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7   A61B5/05

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED
Minimum documentation searched (classification system followed by classification symbols)
IPC 7   A61B

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EPO-Internal, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 5 810 742 A (PEARLMAN ANDREW L) 22 September 1998 (1998-09-22) column 17-18 column 21-24 column 26-27 claims 1-5,10,14; figures 14,15,17,18	21,22
	—/—	

[X] Further documents are listed in the continuation of box C.		[X] Patent family members are listed in annex.

° Special categories of cited documents:

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
21 November 2001	28/11/2001

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31–70) 340–2040, Tx. 31 651 epo nl, Fax: (+31–70) 340–3016	Bernas, Y

Form PCT/ISA/210 (second sheet) (July 1992)

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C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT		
Category *	Citation of document, with indication,where appropriate, of the relevant passages	Relevant to claim No.
A	BROWN B H ET AL: "SIMULTANEOUS DISPLAY OF LUNG VENTIALTION AND PERFUSION ON A REAL-TIME EIT SYSTEM" PROCEEDINGS OF THE ANNUAL INTERNATIONAL CONFERENCE OF THE ENGINEERING IN MEDICINE AND BIOLOGY SOCIETY. PARIS, OCT. 29 - NOV. 1, 1992, NEW YORK, IEEE, US, vol. 5 CONF. 14, 29 October 1992 (1992-10-29), pages 1710-1711, XP000514395 ISBN: 0-7803-0786-0 the whole document	23-26,28
A	US 5 311 878 A (BROWN BRIAN H ET AL) 17 May 1994 (1994-05-17) claim 3; figure 1	23-25
A	EP 1 000 580 A (SIEMENS ELEMA AB) 17 May 2000 (2000-05-17) claims 1-5; figure 1	21
A	DIJKSTRA A M ET AL: "REVIEW CLINICAL APPLICATIONS OF ELECTRICAL IMPEDANCE TOMOGRAPHY" JOURNAL OF MEDICAL ENGINEERING & TECHNOLOGY, NASINGSTOKE, HANTS, GB, vol. 17, no. 3, May 1993 (1993-05), pages 89-98, XP000700151 the whole document	21

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INTERNATIONAL SEARCH REPORT

Information on patent family members

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